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(54) Thermographic apparatus for measuring the temperature distribution in a substantially dielectric medium.

(57) A thermographic apparatus is provided for measuring the temperature distribution in a substantially dielectric medium by detecting the microwave energy emerging from the medium. The apparatus is particularly useful for detecting carcinomas of the human body. The apparatus comprises an array of microwave antennas positioned adjacent the patient's body and electronic means for processing the broad band signals induced in the microwave antennas to determine the temperature prevailing in adjacent volume elements of the patient's body. The electronic means is laid out to process the signals induced in each of the antennas and to correlate each signal with the signal from one of the antennas in a fixed reference location. The correlation produces first and second values containing information relating to the amplitude and phase of the signal received in the or each detecting position of that antenna relative to the amplitude and phase of the signal received by the reference antenna. A computer is used to carry out an inverse transformation of the relative amplitude and phase signals from all the antennas to produce a matrix of temperature values relating to adjacent volumes of the patient's body.

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Thermographic Apparatus for Measuring the
Temperature Distribution in a Substantially
Dielectric Medium

The present invention relates to thermographic apparatus for measuring the temperature distribution in a substantially dielectric medium by detecting the microwave energy
20 emerging from the medium, and has particular reference to thermographic apparatus for use in medical applications. Such medical applications include the detection of carcinomas or lesions of the human body which manifest themselves by a locally elevated temperature and increase in emission
25 of black-body radiation. In this case the human body forms the substantially dielectric medium. The thermographic apparatus can however also be used to determine temperature distributions in other dielectric media.

30 So far as the medical applications are concerned it is known that the region surrounding a carcinoma usually shows an excess of temperature over the mean temperature of the human body, and that this effect is extended to a larger region by the flow of blood from the excess of
35 blood vessels found around the carcinomas. This means that one can in practice detect the carcinomas with a local temperature excess of only 1.5° K using thermographic

1 apparatus. As the temperature distribution is measured
solely by detecting the microwave energy emerging from the
patient's body the techniques of thermography are passive
and non-invasive and can thus be used for the routine screen-
5 ing of people for cancer or other infections without any
danger of side effects. Existing thermographic apparatus
has been successfully used for the detection of carcinomas,
in particular carcinomas of the female breasts, it is not
however really suitable for mass screening because of
10 the large time required to take the necessary measurements.

Typical prior art thermographic apparatus and methods are
described in the articles "Detection of Breast Cancer
by Microwave Radiometry" by A.H. Barrett, P.C. Myers in
15 the journal "Radio Science 12" (supplement) 1977, and
"Centimeter- and Millimeter-Wave Thermography - A survey
on Tumor Detection" by J. Edrich in "The Journal of
Microwave Power" vol. 13, No. 2, 1979, pages 95 to 104.
In the first mentioned article a rectangular waveguide
20 antenna filled with a low loss solid with a dielectric
constant equal to 11 is placed flush against the patient's
skin and a plurality of readings are taken at different
locations, typically nine different locations on each
breast. In each position the antenna receives signals from
25 a substantial volume of the patient's body and a relative-
ly high reading indicates that a carcinoma may be present
in that particular volume of the patient's body. The
carcinoma can be more precisely localised by comparing the
readings from adjacent positions of the antenna. It is
30 possible, although not discussed by these authors, to use an
array of independent antennas to give several simultaneous
readings. This would shorten the time taken for an examin-
ation, but would not give any increase in spatial resolution.
In the method described in the second article the patient
35 is laid on a table and a receiving arrangement, comprising
a reflector and a horn antenna mounted in front of the
reflector, is mounted above the patient and is movable

1 within a cartesian coordinate system. In this way the
microwave radiation emerging from the patient's body is
directed by the reflector to the horn antenna and different
areas of the body can be scanned by moving the receiver
5 arrangement to different coordinate positions. Again a
plurality of readings have to be taken and this takes a
substantial amount of time.

Another method of thermography is described in the article
10 "New Correlation Radiometer for Microwave Thermography"
by A. Mamouni, J.C. van de Velde and Y. Leroy in Electronic
Letters August 6th, 1981, vol. 17, No. 16. This article
describes a receiver which does not consist of a single
probe (or antenna) as in the usual microwave radiometers,
15 but consists of a combination of two or several probes
adjacent to each other and in contact with the patient's
body. The idea underlying this arrangement is that each
probe will receive microwave signals from an adjacent
volume of tissue and that two adjacent volumes of tissue
20 associated with two adjacent probes will have a region of
overlap common to both probes. The electronic circuitry
associated with the two probes is said to be such that the
contribution of the thermal noise generated in the common
volume of overlap can be identified. It is stated that
25 a delay introduced into one of the two signal paths from the
two probes can be used to modify the contribution of the
different sub-volumes in the volume of overlap. The problem
with this arrangement however is that it is only intended
to produce signals from the regions of overlap between ad-
30 jacent probes which will be small, giving better resolution
than with single probes, but temperature mapping of a whole
area of a body is again only possible by producing relative
movement between the probes and the body. Thus again a
plurality of readings is necessary, which takes an un-
35 desirably long time. Furthermore, the spatial resolution
of this and other prior art thermographic apparatus are
relatively restricted.

1 The principal object underlying the present invention is
to provide thermographic apparatus which is able to
give an image of a whole area of a dielectric medium, in
particular of a patient's body, and to measure the tempera-
5 ture distribution in the medium with substantially improved
spatial and thermal resolution in a short time.

In order to satisfy this object there is provided, in
accordance with the invention, and starting from the prior
10 art apparatus described in the article in Electronics
Letters as cited above, thermographic apparatus for
measuring the temperature distribution in a substantially
dielectric medium by detecting the microwave energy
emerging from the medium, the apparatus comprising an
15 array of microwave antennas positioned adjacent the dielec-
tric medium and electronic means for processing the broad-
band signals induced in the microwave antennas in a
plurality of detecting positions to determine the temper-
ature prevailing in a volume element of the dielectric
20 medium, characterised in that the electronic means com-
prises means for processing the signal induced in each
of the antennas in each of the detecting positions and for
correlating each signal with the signal from at least one
of said antennas in a fixed reference location to produce,
25 for the or each detecting position of each antenna, first
and second values containing information relating to the
amplitude and phase of the signal received in that de-
tecting position relative to the amplitude and phase of the
signal received by at least one of the reference antennas;
30 and computing means for forming an inverse transformation of
the first and second values associated with each antenna of
the array, said inverse transformation being a matrix of
temperature values relating to adjacent volumes of said
dielectric medium.

35

This apparatus is thus based on the principle of aperture
synthesis as used in radio astronomy from which it is known

1 that the distribution of amplitude and phase over an antenna's aperture is the Fourier transform of the source's brightness distribution. This is a significant extension of correlation techniques to give an image of the brightness
5 distribution in the whole of the region of response of the individual antennas, which are made so that a large region of the patient's body is in each and every antenna's response pattern. Using the apparatus of the invention it should be possible to measure the temperature distribution
10 over a whole region of a patient's body of approximately the same size as the array size within a very short space of time and with high spatial and thermal resolution. The short period of time required for the measurement not only makes the apparatus suitable for mass screening techniques,
15 but also for monitoring the effects of treatments to the patient on virtually any time scale. Since the signals are correlated across the extremities of the array, the apparatus will have the high resolution appropriate to the array size over the whole region examined. It is believed
20 that the presently proposed apparatus will be capable of detecting deep-seated hot spots 1.5° K above ambient temperature with a positional accuracy of a few millimeters in a short time.

25 The inverse transformation of the first and second values is preferably an inverse Fourier transformation. The inverse Fourier transformation is however theoretically only valid for temperature sources which are at a substantial distance from the antennas. If a straightforward inverse Fourier
30 transformation produces results which are obviously inconsistent it may be necessary to make certain corrections, or to resort to an inverse Fresnel transformation, or even to make corrections and use an inverse Fresnel transformation.

35

In view of this, one embodiment of the apparatus is characterised by the provision of phase adjustment means

1 for adjusting the relative phase of each signal induced in
each of the antennas in the or each detecting position,
relative to the phase of the signal received by at least one
of the reference antennas, prior to effecting the inverse
5 transformation, whereby to compensate for near field
effects. Thus this embodiment envisages that the corrections
required to the inverse Fourier transformation, or possibly
to the inverse Fresnel transformation can be made by correct-
ing the relative phases of the signals prior to effecting
10 the inverse transformation. This adjustment of the relative
phases is most effectively made by software embodied in the
computing means which will enable phase corrections to be
effected and the output to be analysed. If at all possible
it is preferable to use an inverse Fourier transformation
15 rather than an inverse Fresnel transformation because the
calculations involved in performing an inverse Fourier
transformation are substantially easier.

A particularly preferred development of the apparatus is
20 characterised in that an inert dielectric layer of high
permittivity, preferably greater than 10, and of low loss,
preferably better than 5 %, is placed between said array
and said medium. The inert dielectric layer is preferably
greater than 5 cm thick and should preferably be con-
25 toured so that it at least approximately fits the shape
of the patient's body, in order to avoid a skin/air inter-
face which would result in substantial and undesirable
losses. In this way the array of antennas can also be
spaced further away from the patient without significant
30 loss in signal strength. Because the array is spaced
further from the patient the inverse transformation may be
simplified so that it is straightforwardly possible to use
an inverse Fourier transformation.

35 In a further development of the apparatus each of the
antennas may be replaced by two or more antenna elements,
whereby separate polarisation components can be measured

1 either prior to or after the inverse transformation, and means may be provided for compensating for polarisation differences.

5 This embodiment recognises that it may be essential to take account of the state of polarisation of the signals received by the individual antennas.

In a particularly preferred embodiment means are provided
10 for varying the centre frequency and/or the bandwidth at which the measurement of the temperature distribution is carried out.

This embodiment is based on the recognition that the
15 correct choice of frequency and/or bandwidth can yield more information on the temperature distribution inside the patient's body. In particular, the taking of several measurements at different centre frequencies may be necessary for a depth dimension to be added to the measured
20 temperature distribution. It will be appreciated that the measured temperature distribution may be regarded as a two-dimensional projected temperature distribution, similar to an X-ray which is a two-dimensional projected density distribution. A theoretical discussion of the points to be
25 considered when selecting frequency and bandwidth is included at the end of this specification.

In one practical embodiment for varying said centre frequency said electronic means comprises a local oscillat-
30 or and a mixer for generating a difference frequency, with the local oscillator frequency being variable whereby to vary said centre frequency. This embodiment may include amplifier and/or filter means for varying said bandwidth.

35 In an alternative arrangement the electronic means does not include a local oscillator but is characterised in that the centre frequency and bandwidth are both determined by an amplifier and/or a filter.

1 In a further practical embodiment the array of microwave
antennas is spatially fixed, so that each antenna has one
detecting position, and switching means is provided for
connecting said antennas groupwise in turn to said
5 electronic means. In this way it is possible to provide
a large number of antennas in the array without having to
provide a separate electronic channel for each individual
antenna, which would be very costly. Instead selected
groups of antennas are connected in turn to a like number
10 of electronic channels.

Alternatively, a smaller number of antennas is arranged in
an array and means can be provided for moving the array
so that each antenna, other than the reference antenna or
15 antennas, adopts or moves through a plurality of detecting
positions. In this embodiment a respective electronic
channel is associated with each antenna, however the total
number of antennas is substantially reduced.

20 The movable array can be a linear array, which may be
rotated or moved laterally, and which is relatively simple
to construct, it can however also comprise a two-dimension-
al array of non-uniformly spaced antennas. This latter
arrangement is particularly advantageous because the lay-
25 out of the individual antennas in the array can be selected
so that cross-talk between the individual antennas is avoid-
ed.

Said plurality of detecting positions preferably fill the
30 area over which said array is moved. This embodiment recog-
nises that the amount of information which can be gathered
is greatest if, in the course of taking the measurements,
a measurement is made at each element of the area over
which the said array is moved.

35

In a further development of the apparatus intermediate
storage means is provided for temporarily storing said

1 signals from each of said antennas in the or each detecting position. This embodiment makes it possible to temporarily store the signals so that they may be subsequently fed to the computing means in the correct time sequence.

5

In a particularly preferred embodiment the electronic means includes a noise calibration system with a noise generating source, such as a noise diode, adapted to feed noise to each of said antennas and said calibration system

10 is optionally adapted to calibrate the apparatus to take account of reflections at the surface of the medium and/or at boundaries within the medium.

The noise calibration system enables drift in the electronics to be recognised and makes it possible for suitable corrective measures to be taken. In a further development automatic gain control means are provided, being optionally controlled in dependence on a signal derived from the said noise calibration system.

20

In a particularly preferred embodiment the electronic means comprises first multiplier means for forming the products of the signal from at least one of the reference antennas with the signal induced in each of the remaining antennas in the or each detecting position to yield said first values; phase shifting means for shifting the phase of the signal induced in each of the remaining antennas in the or each detecting position by 90° to produce phase shifted signals; and second multiplying means for forming the products of the signal from the same reference antenna with each of the phase shifted signals to yield said second values. When using this embodiment the first and second values have the general form of the product of an amplitude term and the sine and cosine of a phase difference term respectively.

35 The phase and amplitude of the signal received in each antenna in the or each detecting position relative to the signal received by the reference antenna can be directly calculated from the first and second values.

- 1 The antennas preferably comprise dipole antennas or antennas formed as printed circuits on dielectric material as all of the region in the patient being examined must be in the primary beam of each antenna at all times during the
5 examination. Such antennas have the advantage of being both inexpensive and effective and a second, crossed dipole can be easily added to each dipole antenna for polarisation measurements.
- 10 Finally, the present invention recognises that existing hyperthermia apparatus, i.e. apparatus used to treat carcinomas by heating them to elevated temperatures, could be dramatically improved by combining it with the presently proposed thermographic apparatus. Such combined apparatus
15 will be characterised by means for operating the thermographic apparatus and the hyperthermia apparatus in alternative time periods, optionally using the same antennas, whereby to detect the temperature distribution produced by the hyperthermia apparatus and, optionally, by means for
20 controlling the hyperthermia apparatus using the signals from the thermographic apparatus.

Because the thermographic apparatus of the present invention should produce, for the first time, a very rapid
25 read-out of the temperature distribution within the patient's body it should be possible to energise the hyperthermia apparatus for a short while, to measure the actual temperature distribution produced using the thermographic apparatus, to correct the relative phases and
30 amplitudes of the microwave signals fed to the antennas of the hyperthermia apparatus, to carry out further heating to re-measure the temperature distribution, and to repeat this process so as to ultimately produce a desired temperature distribution in the body using the hyperthermia
35 apparatus.

1 In other words the apparatus of the present invention
could be used to provide a real time image of the temperat-
ure distribution around a site being subjected to hyper-
thermia treatment, with the image scanning process being
5 time-shared with the application of hyperthermia power.
With this arrangement it will of course be necessary to
provide adequate thermal protection for the receiver
preamplifier stages. If the time sharing intervals are
small, say 0.1. seconds, the cooling effects of the bio-
10 heat transport mechanisms around the area will be negli-
gible, thus the thermogram represents the true temperature
distribution during the hyperthermia process. This should
greatly aid the preferential destruction of the cancers and
the implementation of high resolution phased array hyper-
15 thermia systems. The apparatus of the present invention
should give a high enough resolution to match the size of
area warmed by the hyperthermia treatment.

A further technique which might be useful is to shine a
20 narrow band noise source thorough the body. The power
required for an adequate signal is quite small (1 mW) and does
not represent a health hazard. This signal would be blinked
and detected synchronously in the computer software. The
method may reveal variations in the opacity of the tissue
25 as a function of frequency which can be detected by image
comparison and may be found in clinical practice to
correlate with abnormalities.

Embodiments of the invention will now be described in
30 further detail by way of example only and with reference
to the accompanying drawings which show:

Fig. 1 a table adapted to support a patient during a
thermographic investigation,
35

Fig. 2 a plan view of a fixed square antenna array,

Fig. 3a a plan view of rotatable linear antenna array,

1 Fig. 3b a plan view of a laterally movable linear antenna
array,

Fig. 4 a schematic section on the line IV-IV of the array
5 of Fig. 3a illustrating the position of a hot spot
which is to be detected,

Fig. 5 a schematic block diagram of the arrangement of
the electronic circuitry used to process the
10 signals from the antennas of the array of Figs. 3
and 4,

Fig. 6 a schematic block diagram illustrating in more
detail the construction of the correlation blocks
15 shown in the block diagram of Fig. 5,

Fig. 7 a diagram illustrating the noise calibration
principle underlying the circuitry of Figs. 5 and 6,
and
20

Fig. 8 the temperature difference seen just into a di-
electric for:
a) an isolated hot spot of diameter 1 cm with
 $T = 1.5^{\circ} \text{K}$ above normal body temperature as seen
25 through various thicknesses of muscles,
b) a layer of warmer material seen under 2 mm of
skin and 1 cm of fat for various thicknesses of
muscle for air ($\epsilon_r = 1$) and dielectric
($\epsilon_r = 30$). In both cases the material has the
30 electrical properties of muscle.

Turning firstly to Fig. 1 there can be seen a table 10
supporting a slab of dielectric 11 of high relative
permittivity $\epsilon_r = 30$ and of low loss (less than 5%). A
35 material of this kind is sold under the trademark "HiK" by
Emerson and Cuming, Canton, Massachusetts 02021, USA. The
upper surface of the slab of dielectric is provided with

1 moulded contours 12 which correspond approximately to the
shape of the human body. Alternatively, slabs of moulded
material could be slidably arranged on the basic slab of
dielectric 11 so that they could be moved to provide a best
5 fit for any particular patient. An antenna array 13 is
positioned immediately beneath the slab of dielectric 11 in
contact therewith and can be slid to and fro beneath the
table on a pair of horizontal rails 14, 15, so that it can
be positioned under the part of the patient's body under
10 investigation.

The antenna array shown in Fig. 1 is in fact a rotatable
linear antenna array as shown in Fig. 3a) and is rotated
by a motor 16 disposed centrally beneath the array. The
15 drive leads for the motor and the connections between the
electronics mounted on the antenna array and the further
electronics are schematically illustrated by the reference
numeral 17. The antenna array may alternatively be a
fixed antenna array such as the square antenna array 13'
20 shown in Fig. 2.

In the square antenna array of Fig. 2 there are eight rows
and eight columns of antennas 18,19 each in the form of a dipole
antenna with one of the antennas 18 forming a reference
25 antenna. Although it would be theoretically possible to
derive signals from each of said antennas simultaneously
this would in fact require 64 processing channels which
would probably be prohibitively expensive. For this reason
it is preferred to provide say eight channels and to
30 use electronic switching techniques to switch between
selected groups of eight antennas. One of the antennas, the
reference antenna 18, must of course always be connected
to one of the channels, the reference channel, because the
existence of a reference antenna is crucial to the meas-
35 urement process as will later become clear from the dis-
cussions of Figs. 5 and 6.

1 The groups of antennas which are connected to the eight
channels at any time are preferably spread across the sur-
face of the array so that cross-talk between individual
antennas is minimised. This is illustrated in Fig. 2 which
5 only shows antennas in eight distinct fields of the array,
namely the antenna 18, which is the reference antenna, and
the further antennas 19 which constitute one of the groups
of antennas which are connected to the remaining seven
electronic channels. It will be understood that identical
10 antennas are provided in each field of the array, these
antennas have however been omitted in order to simplify
the illustration.

Rather than providing such a large number of individual
15 antennas it is also possible to provide fewer antennas
and to move the array to different detection positions as
illustrated in Fig. 3a + b. In this arrangement it is how-
ever essential that one antenna, in this case the antenna
18, should be fixedly positioned for comparison purposes.
20 The linear array of Fig. 3 is rotatable about the central
axis 20 through 180° through a plurality of distinct
detecting positions 21. This can be effected by the motor 16
of Fig. 1 which can be either a stepping motor or a con-
tinuously rotating motor. I.e. readings can be taken while
25 the array is stationary or when it is moving.

An alternative arrangement is shown in Fig. 3b in which
two linear arrays are positioned on either side of a central
fixed reference antenna 18 and can be displaced linearly to
30 the left and to the right relative to the central antenna
as illustrated by the arrows V. This embodiment has the
advantage that the antennas do not rotate and thus respond
to the same polarisation components during the course of
the examination.

35

The array of Fig. 3a is shown again in cross-section in
Fig. 4 which also shows the slab of dielectric 11 and a hot

1 spot 22 which is assumed to lie somewhere in the patient's
body.

Fig. 4 also shows radiation emitted from one point in the
5 hot spot which propagates in the general direction of the
lines 23 and 24 and is detected by the two antennas 18
and 19. It can easily be seen that the signal induced in
the antenna 18 will have a different phase from that of the
signal induced in the antenna 19. Of course the antenna 18
10 and the antenna 19 are not just receiving radiation from the
hot spot 22 but also from every other point in the patient's
body such as the point 22'.

The antenna's beam pattern (its response to radiation from
15 angles θ off its principal axis) must be broad enough to
ensure that it covers all the area to be examined. That
is to say that the whole of the area to be examined should
be in the primary beam of every antenna at all times.

20 It can be seen from Fig. 4 that the paths 23 and 24 will be
mainly in the lossy medium of the patient's body when the
hot spot 22 is at the edge of the area examined. This may
degrade the signal-to-noise ratio so an optional solution
is to take another antenna 18' and use that as a secondary
25 reference antenna. The output from this antenna must also
be combined with the output from the other antennas in the
manner described below to give secondary desired first and
second values. These secondary values can then, in principle,
be eliminated during processing prior to the inverse
30 transformation. The use of this technique does not change
the principles underlying the following discussion.

The signals generated at the antennas 18 and 19 from the
radiation emerging from the hot spot 22 may be described
35 by formulae of the form:

1

$$V_{18} = V_{22} L_{23} f(\theta_{18}) \cos \left(\omega t - \frac{2\pi a}{\lambda} \right) \quad (A)$$

$$V_{19} = V_{22} L_{24} f(\theta_{19}) \cos \left(\omega t - \frac{2\pi b}{\lambda} \right) \quad (B)$$

where V_{22} is the signal leaving the point 22;
 L_{23} and L_{24} are the losses along the paths 23
 10 and 24 of Fig. 4 respectively;
 $f(\theta)$ is the beam pattern of the individual
 antennas (assumed to be the same for all anten-
 nas), with the angles θ_{18} , θ_{19} being defined as
 shown in Fig. 4; and ω is the angular frequency
 15 of the radiation leaving the point 22;
 a and b are the lengths of the paths 23 and 24
 respectively, and λ is the wavelength in the
 dielectric.

The expressions for the signals arising at the antennas 18
 20 and 19 from the point 24 are similar vis:

$$V'_{18} = V'_{22} L_{23} f(\theta'_{18}) \cos \left(\omega t - \frac{2\pi a'}{\lambda} \right) \quad (C)$$

$$V'_{19} = V'_{22} L_{24} f(\theta'_{19}) \cos \left(\omega t - \frac{2\pi b'}{\lambda} \right) \quad (D)$$

and indeed similar equations can be written for the energy
 arriving at the two antennas from each point in the
 30 patient's body.

On summing these signals one obtains

$$\langle V_{18} \rangle = \iint V(\theta, a) \cdot L(\theta, a) \cdot f(\theta) \cdot \cos \left(\omega t - \frac{2\pi a}{\lambda} \right) \cdot d\theta da \quad (E)$$

$$\langle V_{19} \rangle = \iint V(\theta, b) \cdot L(\theta, b) \cdot f(\theta) \cdot \cos \left(\omega t - \frac{2\pi b}{\lambda} \right) \cdot d\theta db \quad (F)$$

- 1 These can be expressed in the form $\langle V_{18} \rangle = \tilde{V}_{18} \cos$
 $(\omega t - \varphi_{18})$ and $\langle V_{19} \rangle = \tilde{V}_{19} \cos(\omega t - \varphi_{19})$ where \tilde{V}_{18} and
 \tilde{V}_{19} are mean voltages and φ_{18} and φ_{19} are mean phase
shifts and these signals now have to be processed and
5 correlated to obtain the amplitude and phase of the net
signal received at antenna 19 relative to the signal received
at antenna 18. This processing is done by the electronic
circuitry shown in Figs. 5 and 6.
- 10 Referring firstly to Fig. 5 it can be seen that the signals
received by each of the antennas is directed through a
respective channel. Each channel first contains a circulator
25 which is a device for ensuring that energy flows
primarily from the antenna to the detection channel and
15 not vice versa. Each circulator 25 is followed by a band-
pass filter 26 which may be incorporated in the preamplifiers
27.

Thereafter the signals in each channel are amplified in
20 preamplifiers 27 and are passed to image rejecting circuits
28 which reduce the noise present in the mixer image band,
coming from the broadband preamplifiers. These filters
reduce the uncorrelated noise present in the signals thus
improving the signal-to-noise ratio. At this stage the
25 signals are fed into respective single-sideband mixers 29
and are mixed with a signal supplied by a local oscillator
30, which is common to all electronic channels. The
difference frequencies produced by the mixers 29 are then
fed to low pass filters 29' and then to the respective
30 frequency amplifiers 31. The signal in the reference
channel associated with the reference antenna 18 is then
passed through a phase switching device 32 which will be
described later. Thereafter the signal from the reference
channel is passed to the correlators 33 provided in the
35 remaining channels. One of these correlators is illustrated
in detail in Fig. 6.

1 As can be seen the signal received by the correlator 33
 of Fig. 6 from the reference channel is first split at a
 T-splitter 34 into two signals which are passed to res-
 pective inputs 35 and 36 of first and second linear multi-
 5 pliers 37, 38 with a dynamic range greater than 40 dB.

The signal coming from the respectively associated antenna
 19 is passed to a broad band 90° splitter 39 (bandwidth
 500 MHz on a 3 GHz system) which has two outputs 40 and 41.
 10 The signal at the output 40 has not been phase-shifted
 and is fed to a second input 42 of the first multiplier 37.
 The second output 41 has a phase shift of 90° relative to
 the input signal and is passed to the second input 43 of
 the second linear multiplier 38. The signals received at
 15 the inputs 35 and 42 have the general form

$$V_{35} = \tilde{V}_{18} \cos(\omega t - \varphi_{18}) \quad (G)$$

$$V_{42} = \tilde{V}_{19} \cos(\omega t - \varphi_{19}) \quad (H)$$

20

and the signals received at the inputs 36 and 43 have the
 general form

$$V_{36} = \tilde{V}_{18} \cos(\omega t - \varphi_{18}) \quad (I)$$

25

$$V_{43} = \tilde{V}_{19} \cos(\omega t - \varphi_{19} + \frac{\pi}{2}) \quad (J)$$

The signals appearing at the outputs 44 and 45 of the first
 and second linear multipliers 37 and 38 respectively have
 30 the general form:

$$S_{44} = \overline{V_{18} V_{19}} \cos(\varphi_{18} - \varphi_{19}) \quad (K)$$

$$S_{45} = \overline{V_{18} V_{19}} \sin(\varphi_{18} - \varphi_{19}) \quad (L)$$

35

The bar denotes the time average of the product of the
 signals \tilde{V}_{18} and \tilde{V}_{19} .

1 These signals are now integrated by converting them via
 respective voltage to frequency convertors 46, 47 into
 frequency signals which are then integrated in 16 bit
 counters 48, 49 and fed onto a 16 bit databus via respect-
 5 ive buffer stores 50, 51. A cycle of 8 such 20 msec samples
 labelled A - H is given in the example shown in Fig. 7.
 Each sample represents the correlated power received during
 the sample period. The sign of the correlation is reversed for
 each alternate sample by the phase switch 32, shown in
 10 Figs. 5 and 6. The samples are combined by subtracting in
 pairs thus (A-B), (C-D), (E-F), (G-H) for the example
 given. This procedure effectively corrects for zero drifts
 in the multipliers 37, 38 and the voltage to frequency
 convertors 46, 47. The cosine correlation is thus formed
 15 using elements 37, 46, 48 and 50 to form $A_c, B_c, C_c, D_c,$
 E_c, F_c, G_c, H_c and elements 38, 47, 49 and 51 to form $A_s,$
 $B_s, C_s, D_s, E_s, F_s, G_s, H_s$. These are fed by the 16 bit
 databus 52 into a microprocessor which forms the com-
 puting means and which is adapted to manipulate the first
 20 and second values received from the buffer stores 50, 51
 via the databus to calculate the relative phase and
 amplitude at each antenna 19 relative to the reference
 antenna 18. The relative phase and amplitude are calcul-
 ated from the equations

25

$$\text{Cosine} = (A_c - B_c) + (C_c - D_c) + (E_c - F_c) + (G_c - H_c) = S_{44}, \quad (M)$$

$$\text{Sine} = (A_s - B_s) + (C_s - D_s) + (E_s - F_s) + (G_s - H_s) = S_{45}, \quad (N)$$

$$30 \text{ Then relative amplitude} = A = \sqrt{S_{44}^2 + S_{45}^2} \quad (O)$$

$$\text{relative phase} = \phi = \tan^{-1}(S_{45}/S_{44}) \quad (P)$$

These are computed for each correlated antenna pair.

35

The microprocessor is then programmed to calculate the
 inverse Fourier transform from the relative phase and
 amplitude values for each of the antennas in each of the

- 1 detection positions, and to produce the output as a matrix of temperature values relating to adjacent volumes of said dielectric medium. This matrix of temperature values can of course be displayed in various ways, for example on the screen of a monitor, possibly with different colours being ascribed to different temperature values, and can also be stored on a floppy disk for future reference or for comparison with images made during successive examinations.
- 10 If, as previously mentioned, it is found that the Fourier transformation is not satisfactory because the array is too close to the patient then it is possible to change the relative phases calculated from the first and second values by calculated or previously empirically derived amounts and to improve the final result of the inverse transformation.

Alternatively it is possible to calibrate the apparatus by using a suitable body of dielectric with known high temperature hot spots therein. The phase corrections required to bring the measured distribution calculated by the inverse Fourier transformation into agreement with the actual temperature distribution can be recorded and subsequently used to correct the relative phases calculated when using the apparatus on a patient.

So far as carrying out the inverse Fourier transformation is concerned the microprocessor is programmed to carry out the following calculation:

30

$$f(l,m) = \iint g(x,y) \exp \left\{ 2\pi i (xl+ym) \right\} dx dy$$

In this expression l and m are the coordinates of a plane close to the patient's body and the function $f(l,m)$ is the desired matrix of temperature values. For any given value of l and m , $f(l,m)$ gives the value of the temperature contribution through the body projected onto this plane. x and y

1 are the coordinates, expressed in wavelengths, of the plane
 containing the antenna array. The origin of this coordinate
 system, i.e. $x = 0$, $y = 0$ is preferably the reference
 antenna and the origin of the plane close to the patient's
 5 body, i.e. $l=0$, $m=0$ will be directly over this. The
 function $g(x,y)$ is complex and is the measured distribution
 across the plane of the array. As an example for the
 positions in Fig. 4 and taking antenna 18 as the reference
 antenna, the measured value for $g(x,y)$ will be $g(x_{19}, y_{19}) =$
 10 $= S_{44} + i S_{45}$, where $i = \sqrt{-1}$.

When the patient is close to the array a Fourier transform
 will not give adequate results. In this case close to
 means less than the "far field" criterion given by
 15 $z = 2 D^2/\lambda$ using the notation of Fig. 4. When z is only
 slightly less than $2 D^2/\lambda$ it will be possible to obtain
 adequate results by putting in a phase correction to the
 Fourier transform, but for smaller values of z the patient
 is in the near field or Fresnel region. In this case the
 20 microprocessor has to be programmed to carry out an inverse
 Fresnel transform:

$$f(l,m) = \iint \exp \{i(xl + ym)\} g(x,y) \exp \{-i(a_x x^2 + a_y y^2)\} dx dy$$

25 In this equation a_x and a_y are parameters of the form:
 $a_n = k_n \pi / r_n \lambda$ where r is the distance from the reference
 antenna to the point in the l,m plane and r_n is the length
 of the projection of r on the appropriate axis. k_n is
 a parameter which will depend on the phase shifts along
 30 the path a_n . The other symbols have their previous meanings.
 This equation gives the transform for a two dimensional
 picture of the temperature distribution, analogous to an
 x-ray picture. It may be possible in practice to obtain
 some estimate of the temperature distribution in the z
 35 direction, in which case the equation will need to be
 modified appropriately.

1 One of the problems which can arise with thermographic
apparatus is the problem of drift which can seriously
affect the results. In order to overcome this drift, or at
least to recognise it so that its effects may be taken
5 into account, the apparatus may be provided with automatic
gain control circuits which may be incorporated as a modification to the amplifiers 31. The present apparatus is
provided with a continuous noise calibration system as a
preferred alternative solution.

10

This system consists essentially of the noise diode 54
illustrated in Fig. 5 which feeds a small noise signal into
each of the antennas 18, 19. A circuit 55 is provided for
modulating the output of the noise diode 54 with a square
15 wave at a frequency F . The result of this is equivalent to
turning the noise diode on and off at the frequency F ,
the noise diode should however not actually be switched
on and off as this would make it unstable.

20 The phase switch 32 present only in the electronic channel
for the reference antenna serves to periodically change the
phase of the signal in the reference channel by 180° and at
a frequency $2F$ which must be an integer multiple or vulgar
fraction of F . The phase switch 32 is driven by a phase
25 switch oscillator 32' which is linked synchronously to the
noise switch oscillator 55' and the correlators 33, or
alternatively (as shown in broken lines) to the noise switch
oscillator 55' and the computer 53. This makes phase
sensitive detection possible.

30

The effect of the noise calibration and of the phase switch
is illustrated by an example shown in Fig. 7a over one
sample period of 160 msec. It will be noted that this
period of 160 msec is subdivided into eight portions each
35 of length $1/4P$. As has been described earlier the circuit
of Fig. 6 performs the correlated integral for each of the
20 msec portions A - H and all eight values from each

1 multiplier are passed onto the databus. The microprocessor,
in addition to computing the sine and cosine correlation
terms using the formula $(A-B)+(C-D)+(E-F)+(G-H)$ computes
also the signal contributed by the noise diode 54 in
5 Fig. 5 and which is demodulated using the formula $(A-B)-$
 $-(C-D)+(E-F)-(G-H)$. This signal in a period of 160 msec
has a significant measurement error due to noise but this
can be reduced to a sufficiently accurate level by fitting
a linear least squares solution to the noise diode cali-
10 bration readings over a period of a minute or so. This
signal can also be used to control the gain of the
appropriate channel (automatic gain control AGC). In any
case it is used in the software to provide a normalisation
calibration factor so that changes in receiver gain can be
15 compensated for in the analysis. As is shown in Fig. 7b
this noise calibration vector must be subtracted from the
outputs of the cosine and sine channel readings to get
true amplitude and phase information.

20 When the reflection correction technique is used the noise
diode is used to inject noise into the antenna lines so that
this signal is radiated out into the dielectric media (at
a very low level). This corrects for reflection losses at
dielectric boundaries. The amplitude and phase of the
25 reflected signal can be demodulated in the same way as
the gain calibration signal, and has also to be subtracted
from the sine and cosine channels to get the true readings.
This noise reflection correction technique has been des-
cribed in U.S. Patent 4,235,107 for single antenna measure-
30 ments.

As mentioned earlier it will probably be necessary to
take account of the state of polarisation of the signals
received in each channel. Polarisation components can be
35 defined and measured in many ways but all methods of
determining them in a synthesis system rely on antenna
elements at each antenna position that each respond to one

1 or both polarisation components. The signal from one element (e.g. 18 in Fig. 6) is then combined with the signal from either its "double" or its "complement" at the other antenna position (19, Fig. 6). The other pair then requires
5 another set of electronics (Figs. 5+6). Full polarisation data is then obtained by either adding or subtracting the outputs of the two pairs either in hardware or in software before or after the transformation. The combination
10 chosen is a design parameter. It may however also be possible to make a combined antenna that responds to more than one type of polarisation giving an output proportional to the total signal arriving at that point.

In order to provide good contact between the patient and
15 the table it may be desirable to use a high dielectric constant cushion which may be capable of being moulded to the patient's body shape or a thin polythene envelope of warm water. In either case the intention is to avoid a patient air interface which would give rise to undesirable
20 losses.

Finally, the following comments are made concerning the system parameters and components.

25 a) Spatial Resolution

Resolution is used here in the sense of the full beam width given by $\theta = \lambda/D$. For an array at a distance z from the object being examined the resolution is then $r = \lambda z/D$ cm,
30 expressed in terms of distance in the patient (see Fig. 4). D/z can only have a value between 0.5 and 5 for practical reasons, with 4 being a reasonable value, but the wavelength can be reduced by a factor of 5.5, using a dielectric with $\epsilon_r = 30$ between the patient and the array giving
35 a corresponding improvement in resolution. This gives a resolution of

1

$$r = \frac{cz}{\nu D \sqrt{\epsilon_r}} \quad (Q)$$

5 where c is the velocity of light, and ν is the observing frequency = $\omega/2\pi$.

r is given in Table 1 (page 33) for several frequencies with $D/z = 4$. The use of a dielectric has two further advantages: (i) the small antennae can be mounted on a rotating piece of similar dielectric enabling strip line techniques to be used in their construction, and perhaps also in the local oscillator supply, and signal amplifier and mixer chains; (ii) suitable shaping of the material to give good contact to the patient's body-shape will considerably improve the matching of the body to the antennae as there is then no longer a skin to air interface (high to low ϵ_r).

20 The resolution given above, which is of the order of 0.5 to 2.5 cm for $3 \text{ GHz} > \nu > 1 \text{ GHz}$, is not the error in the determination of the position of the hotter region, as beam-fitting techniques can give the position of small sources to within 0.1 beam widths even when the signal-to-noise ratio of the signal is only 5:1. In this context beam fitting means numerical techniques which give a best fit of a diffraction pattern to a small part of the matrix of temperature values.

30 b) Operating Frequency

The choice of frequency must consider four factors affecting the visibility of a hot region of fixed size inside the body:

35

- 1 (i) the radiation's penetration (1/e) depth in tissue increases with wavelength, the absorption coefficient is given approximately by the following empirically derived formula:

5

$$\alpha = 2.2 \times 10^{-5} \sqrt{\nu} + 0.1 \quad (R)$$

over the frequency range 100 MHz to 3 GHz;

- 10 (ii) the hot spot's optical depth increases with frequency giving an increase in effective temperature rise above the surroundings;

15

Optical depth (τ) is a measure of the absorption of radiation of a particular wavelength as it passes through a lossy medium. For a piece of this medium, the radiation intensity emerging is $e^{-\tau}$ times the incident radiation intensity. For black-body radiation from the lossy medium, which we consider here, the intensity arising from it will be $(1 - e^{-\tau})T$ where T is the temperature of the medium;

20

- 25 (iii) when the hot-spot's diameter is less than the diameter of the antenna arrays synthesised beam at that point the hot-spot's measured temperature is reduced by the ratio of the beam diameter to hot-spot diameter;

- 30 (iv) the reflections at interfaces inside the body between materials of different permittivities cause losses.

Quantifying effects i, ii and iii, and using equations (A) and (B), lead to a relation for the change in temperature seen at the skin's surface

35

$$\Delta T_e = \Delta T \epsilon_r \left(\frac{D k \sqrt{V}}{zc} \right)^2 \left[1 - \exp \left\{ - (2.2 \times 10^{-5} \sqrt{V} + 0.1) k \right\} \right] \cdot \exp \left\{ - (2.2 \times 10^{-5} \sqrt{V} + 0.1) b \right\} \quad (S)$$

where: ΔT is the hot spot's actual temperature rise above its surroundings, = 1.5 K in this example,

k is the diameter of the hot spot assumed to be a uniform disc of thickness k , = 1 cm in this example,

b is the depth of the hot spot below the body surface.

The value of ΔT_e for several parameters is shown in Fig. 8a. Fat has been ignored here because its absorption coefficient is much lower than above, but shows the same variation with frequency. The effect of mismatches at boundaries due to effect (iv) above can be seen in Fig. 8b which gives the temperature measured just outside the skin for a 1.5 cm thick layer of hotter material seen through 2 mm of skin and layers of muscle and fat. This is based on the following relation which has been mathematically derived from the information given in the paper by J. Edrich in "The Journal Of Microwave Power" vol. 14, No. 2, 1979, pages 95 to 104:

$$\Delta T_e = \Delta T \left[1 - \exp \left(- \frac{z_t}{d_t} \right) \right] \cdot \exp \left(- \frac{z_f}{d_f} \right) \cdot e_{fs} \cdot \exp \left(- \frac{z_s}{d_s} \right) \cdot e_{so} \cdot \exp \left(- \frac{z_m}{d_m} \right) \cdot e_{mf} \quad (T)$$

where: z_t, z_f, z_s, z_m are the thicknesses of the layers of tumour, fat, skin and muscle respectively,
 d_t, d_f, d_s, d_m are the $1/e$ penetration depths of the materials at a given frequency,
 e_{fs}, e_{so}, e_{mf} are the emissivities from fat to skin, skin to outside the body, and muscle to fat.

1 Consideration of Fig. 8 shows that for small hot spots
 higher frequencies are better, until either ΔT_e levels
 off or equation (R) breaks down (> 3 GHz), whilst for
 larger hot areas the lower frequencies are better. There-
 5 fore the choice of operating frequency, within the range
 900 to 3000 MHz, must depend on the particular application
 being considered.

Closely related to the operating frequency is the usable
 10 bandwidth. This must be sufficiently small for the phase
 difference between signals at the antennae with the largest
 separation, A_1 and A_2 in Fig. 4, not to change by more
 than 90° across the band for points at the edge of the
 picture area. Taking the picture area to be the same as
 15 the antenna array area and directly over it, we come to
 the relation:

$$\Delta \nu < \frac{\nu^2}{4c} \frac{xz}{(x-z)} \quad (U)$$

20 where $\Delta \nu$ is the bandwidth and $x = z^2 + D^2$. This is a very
 weak constraint, varying from 11 % to 40 % of the operating
 frequency over the range 915 MHz to 3000 MHz, and the
 bandwidths given on page 33 are less than this.

25 c) The Antenna Array

In the far-field case, the number of antennas is determined
 by the minimum separation between antennas (ΔD) for the
 lowest Fourier component to have a radius larger than the
 30 picture size. In our case, but with far-field approximat-
 ions, the restriction is that $\lambda/\Delta D > \phi$ where ϕ is
 the angle subtended by the examined area at one element
 of the array, with $\phi = \tan^{-1} (D/z)$. Expressed in terms
 of frequency we have:

$$\Delta D < \frac{c}{\nu} \left[\epsilon_r \tan^{-1} \left(\frac{D}{z} \right) \right]^{-1} \quad (V)$$

and the number of antennae.

1

$$n = \frac{D}{\Delta D} + 1 \quad (W)$$

For $D/z = 4$, ΔD is of the order of one or two centimetres
 5 so dipoles, with suitable balancing networks, are ideal
 as the individual antennas. The patient is in the near
 field of the array so it is not possible to reduce the
 number of antennas by the usual procedure of having un-
 equally spaced antennas separated by multiples of ΔD and
 10 reconstructing the whole aperture using combinations of
 these as used in radio astronomy. Although this increases
 the array costs it is offset to some extent by the fact
 that the path compensators and phase rotators used in
 astronomy are unnecessary and the data rate is not too
 15 high for a small computer handle.

d) Preamplifiers

Many preamplifiers are necessary as all antenna pairs are
 20 present so they must be inexpensive. Since the array looks
 at a 300 K patient this is the lowest possible overall
 system noise temperature with a perfect amplifier. With the
 calibration methods discussed in the specification direct-
 ional coupler, circulator and filter are needed between
 25 the dipoles and preamplifiers giving a small degradation
 in system noise temperature. A simplified sketch of the
 system is shown in Fig. 5. For our calculations we have
 assumed a system noise temperature of $600 \text{ K} = 300 \text{ K patient}$
 $+ 300 \text{ K amplifier}$ since such amplifiers are readily
 30 available at low cost.

e) The Correlators and Continuous Calibration

The amplitude (A) and phase (ϕ) are measured as sine (S)
 35 and cosine (C) terms, where $A^2 = S^2 + C^2$ and $\phi = \tan^{-1}(S/C)$,
 by cross correlation of the signal from the central receiver
 with that of a receiver at the appropriate spacing for each

1 of the spacings. In the system proposed the bandwidths are
much larger than those used in astronomical synthesis so
the correlators must be carefully designed so that the
relative phase between inputs is less than a few degrees
5 across the whole bandwidth (200 MHz for the 1.5 GHz system).

In addition they need at least a 40 dB dynamic range
(1 mK to 10 K). This specification can be achieved by the
use of linear multipliers and by 180° broad band phase
10 switching in one of the I.F. inputs (see Fig. 6). When
synchronous demodulation is applied this removes the 5 %
or so square law response inherent in the multipliers at
higher signal levels and simultaneously corrects for zero
drift in the multipliers and following voltage-to-frequency
15 convertors. These convertors, in conjunction with buffered
counters, act as integrators. Gain calibration for each
channel is derived by a modulated noise source switched
at half the phase inversion rate and which is weakly
(-25 dB) but coherently coupled into the input circuit of
20 each preamplifier, see Fig. 5. Software demodulation makes
it possible to carry out real time correction, during the
analysis, for system drifts of phase and amplitude and
provides engineering monitoring of possible system mal-
functions.

25

f) Inverse Transform and Image Matrix

During the rotation of the array we need to collect $D^{\sim}/\Delta D$
30 samples each of N-1 amplitudes and phases to match the mini-
mum separation limit calculated in section c. For the
1.5 GHz system this gives 57 samples each of 18 amplitudes
and phases. We have chosen 64 samples each lasting
160 milliseconds in the software demodulation example previ-
35 ously described. As the array rotates the amplitude and
phase information is interpolated into a matrix of values
in the aperture plane and the continuous calibration

1 corrections previously described are computed and applied
at the end of the scan directly into the aperture plane
array. This array is then transformed to form the image.
This transform may well require special purpose hardware
5 to take full advantage of the short 10 second exposure time
required by the receivers.

The image matrix requires a minimum of two points per
resolution element which for the 1.5 GHz system is 0.91 cm
10 (Table 1, page 27) giving a grid point separation of 0.45 cm.
A 50 cm square image then needs a matrix of 111 x 111
points. If each element has 2 byte accuracy (1 in 65536) we
can provide a temperature range of 10 K relative to some
conveniently chosen zero for the image with an accuracy of
15 0.15 mK due to round-off. Such an image would require 25 K
bytes including identification and calibration information
and sixteen such images could be stored on a single
0.5 Megabyte floppy disk including patient case history
details. This would give a convenient cheap filing
20 method for hospital use, allowing later off-line inspection
by a consultant and eventual transfer to mass storage for
large sample statistical analysis.

The image matrix could well have the format described in
25 the article "NOD 2 A general system of analysis for
radioastronomy" by C.G.T. Haslam in Astronomy and Astro-
physics Supplement vol. 15 p.p. 333-335, 1974 which is
very efficient in computer time when used with two-
dimensional regridding and interpolation routines. These
30 operations will be essential for comparing pictures taken
on different days, since the patient will not lie in
exactly the same place, so superposition, shifts of scale,
translation and rotation will be required before detailed
comparison can be made. Suitable software and necessary
35 algorithms, including techniques for combining over-
lapping images, have been described, in an astronomical
context, in the article "NOD 2 A Generalised System of
Data Analysis for Astronomy which has Application to Image

1 Processing" by C.G.T. Haslam, N.C. Haslam and D.T. Emerson,
Technischer Bericht 56 des Max-Planck-Instituts für
Radioastronomie Bonn, 1980.

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Table 1**System Parameters.**

Maximum antenna separation 50 cms.

Distance from patient to antenna array 12.5 cms.

Dielectric constant of table 30

System noise temperature 300 K + 300 K from patient.

Frequency	915	1500	2450	3000	MHz
Resolution	1.49	0.91	0.55	0.45	cms
Number of antennae in array	12	19	30	37	
Bandwidth	100	200	400	500	MHz
R.m.s. noise in 10 second exposure	5.47	3.07	1.73	1.39	milli Kelvin
Time taken to see 1.5 K warmer layer below 2 mm skin, 1 cm fat and 4 cm muscle	6	9	57	222	seconds

- 1 1. Thermographic apparatus for measuring the temperature
distribution in a substantially dielectric medium by
detecting the microwave energy emerging from the medium,
the apparatus comprising an array (13; 13') of micro-
5 wave antennas (18, 19) positioned adjacent the dielectric
medium and electronic means (Figs. 5 + 6) for processing
the broad-band signals induced in the microwave antennas
(18, 19) in a plurality of detecting positions (21) to
10 determine the temperature prevailing in a volume element
of the dielectric medium, characterised in that the
electronic means (Figs. 5 + 6) comprises means (25, 26,
27, 28, 29, 29', 30, 31, 33) for processing the signals
induced in each of the antennas (18, 19) in each of the
15 detecting positions (21) and for correlating each signal
with the signal from at least one of said antennas (18)
in a fixed reference location to produce, for the or each
detecting position (21) of each antenna (19), first and
second values containing information relating to the
20 amplitude and phase of the signal received in that de-
tecting position relative to the amplitude and phase
of the signal received by at least one of the reference
antennas (18); and computing means (53) for forming an
inverse transformation of the first and second values
25 associated with each antenna of the array, the result of
said inverse transformation being a matrix of temperat-
ure values relating to adjacent volumes of said dielec-
tric medium.
2. Thermographic apparatus in accordance with claim 1 and
30 characterised by the provision of phase adjustment means
for adjusting the phase of each signal induced in each
of the antennas (19) in the or each detecting position,
relative to the phase of the signal received by at least
one of the reference antennas (18), prior to effecting
35 the inverse transformation, whereby to compensate for
near field effects.

- 1 3. Thermographic apparatus in accordance with either of
the preceding claims, characterised in that an inert
dielectric layer (11) of high permittivity, preferably
greater than 10, and of low loss, preferably better than
5 %, is placed between said array (13, 13') and said
medium.
4. Thermographic apparatus in accordance with claim 3,
characterised in that said dielectric layer (11) is
greater than 5 cms thick.
5. Thermographic apparatus in accordance with any one of
the preceding claims, characterised in that each of the
antennas (18, 19) is replaced by two or more antenna
elements whereby separate polarisation components can
be measured either prior to or after the inverse
transformation, and, optionally, in that means are pro-
vided for compensating for polarisation differences.
6. Thermographic apparatus in accordance with any one of
the preceding claims, characterised in that means (29,
29'; 30, 31) are provided for varying the centre
frequency and/or the bandwidth at which the measurement
of the temperature distribution is carried out.
7. Thermographic apparatus in accordance with claim 6,
characterised in that said electronic means (Figs. 5 + 6)
comprises a local oscillator (30) and a mixer (29) for
generating a difference frequency, with said local
oscillator frequency being variable whereby to vary said
centre frequency.
8. Thermographic apparatus in accordance with claim 7,
characterised in that amplifier (31) and/or filter means
is provided for varying said bandwidth.

1 9. Thermographic apparatus in accordance with claim 6
wherein said electronic means does not include a mixer
and/or local oscillator, characterised in that the centre
frequency and bandwidth are determined by an amplifier.

5

10 10. Thermographic apparatus in accordance with any one of
the preceding claims, characterised in that said array
(13') of microwave antennas (18, 19) is spatially fixed,
whereby each antenna (18, 19) has one detecting
position; and in that switching means is provided for
connecting said antennas groupwise in turn to said elec-
tronic means (Figs. 5 + 6).

15 11. Thermographic apparatus in accordance with any one of the
preceding claims 1 to 9, characterised in that means
(16) is provided for moving said array (13) so that each
antenna (19) other than said reference antenna (18) or
antennas (18') adopts or moves through a plurality of
detecting positions.

20

12. Thermographic apparatus in accordance with claim 11,
characterised in that said movable array comprises one
or more linear arrays (13).

25 13. Thermographic apparatus in accordance with claim 11,
characterised in that said movable array comprises a
plurality of non-linearly spaced antennas.

30 14. Thermographic apparatus in accordance with any one of
the preceding claims, characterised in that intermediate
storage means is provided for temporarily storing
said signals from each of said antennas (18, 19) in the
or each detecting position.

35 15. Thermographic apparatus in accordance with any one of
the preceding claims, characterised in that said elec-
tronic means (Figs. 5 + 6) includes a noise calibration

1 system (54, 55, 55') with a noise generating source (54),
such as a noise diode, adapted to feed noise to each of
said antennas (18, 19); and in that said calibration
system (54, 55, 55') is optionally adapted to calibrate
5 the apparatus to take account of reflections at the
surface of the medium and/or at boundaries within the
medium.

16. Thermographic apparatus in accordance with any one of the
10 preceding claims, characterised in that said electronic
means (Figs. 5 + 6) further comprises means for con-
trolling the gain of the said electronic means automatic-
ally, said automatic gain control being optionally con-
trolled in dependence on a signal derived from the said
15 noise calibration system.

17. Thermographic apparatus in accordance with any one of
the preceding claims characterised in that said elec-
tronic means (Figs. 5 + 6) comprises first multiplier
20 means (37) for forming the products of the signal from
the reference antenna (18) with the signal induced in
each of the remaining antennas (19) in the or each
detecting position to yield said first values; phase
shifting means (39) for shifting the phase of the
25 signal induced in each of the remaining antennas in
the or each detecting position by 90° to produce phase
shifted signals; and second multiplying means (38) for
forming the products of the signal from the reference
antenna (18) with each of the phase shifted signals to
30 yield said second values.

18. Thermographic apparatus in accordance with claims 15 and
17 characterised in that said electronic means (Figs. 5
+ 6) further comprises means (32) for periodically
35 changing the phase of the signal from the reference
antenna by 180° with the change taking place at a first
frequency (2F) and optionally further means (55) for

- 1 periodically switching the output of said noise source
 (54) between a selected value and zero at a second
 frequency (F) equivalent to said first frequency (2F)
 divided by or multiplied by an integer, and means for
5 periodically sampling said first and second values at
 intervals of time $(1/4F)$ equivalent to one half of the
 reciprocal of said first frequency; and in that said
 computing means (53) is adapted to process said sampled
 first and second values to derive calibration information
10 and signal information.
19. Thermographic apparatus in accordance with any one of
 the preceding claims characterised in that said antennas
 comprise dipole antennas (18, 19) or antennas formed as
15 printed circuits on dielectric material.
20. The combination of thermographic apparatus in accordance
 with any one of the preceding claims with hyperthermia
 apparatus characterised by means for operating the
20 thermographic apparatus and the hyperthermia apparatus
 in alternative time periods, optionally using the same
 antennas, whereby to detect the temperature distribution
 produced by the hyperthermia apparatus and, optionally,
 by means for controlling the hyperthermia apparatus
25 using the signals from the thermographic apparatus.
21. The combination of thermographic apparatus in accordance
 with any one of the preceding claims with a narrow band
 noise source located on the side of said dielectric
30 medium opposite to said thermographic apparatus, wherein
 said narrow band noise source is blinked and detected
 synchronously by said computing means.

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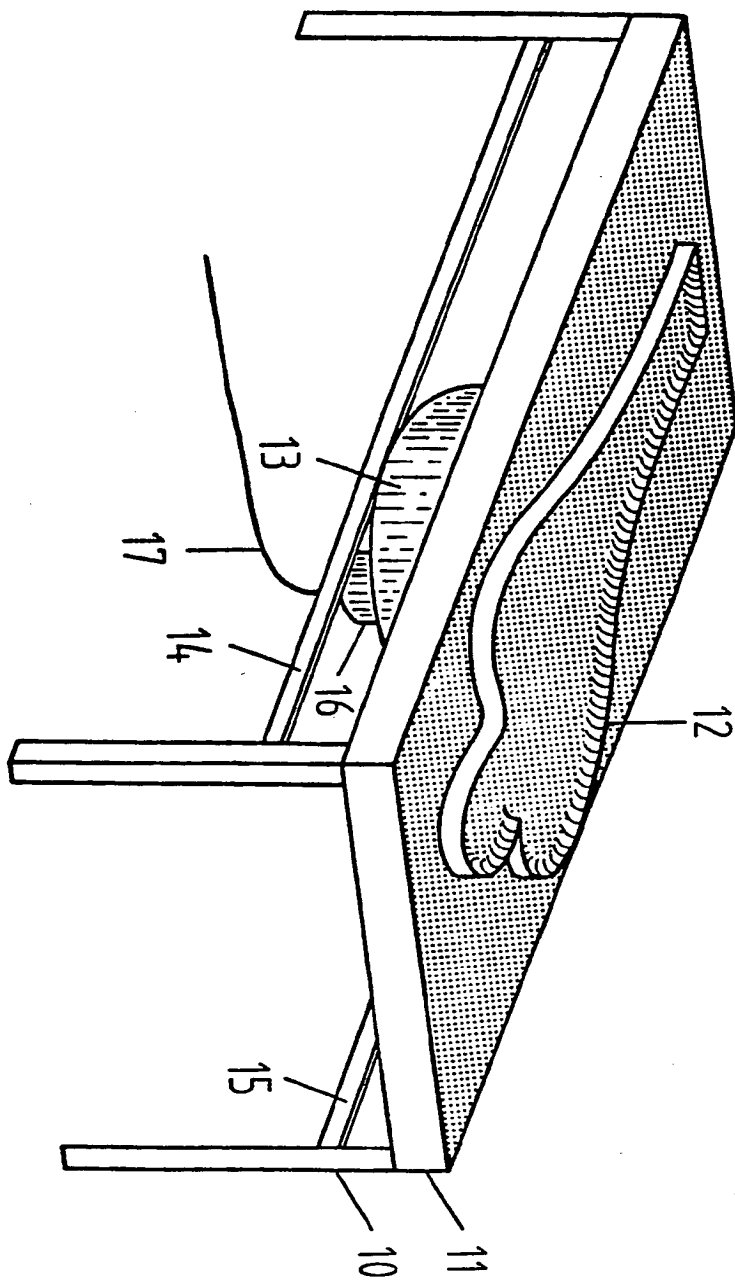


FIG. 1

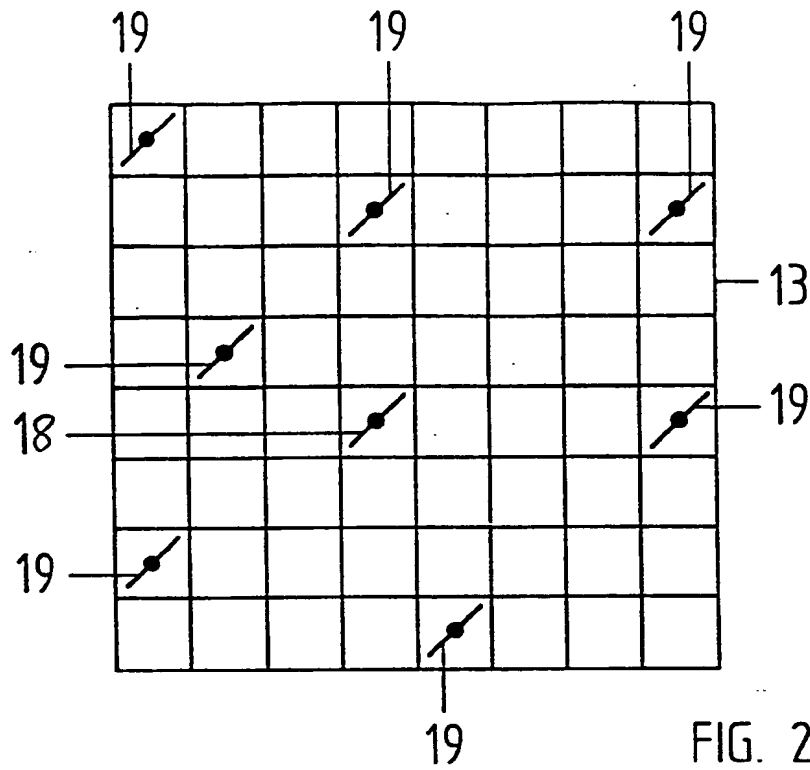


FIG. 2

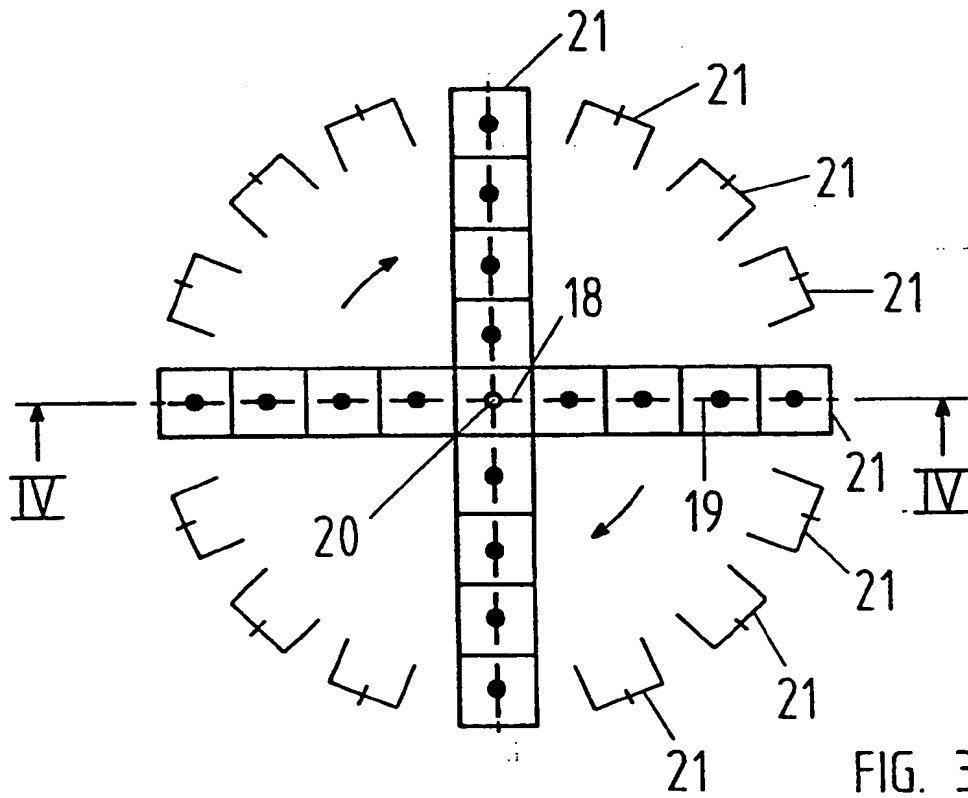


FIG. 3 a

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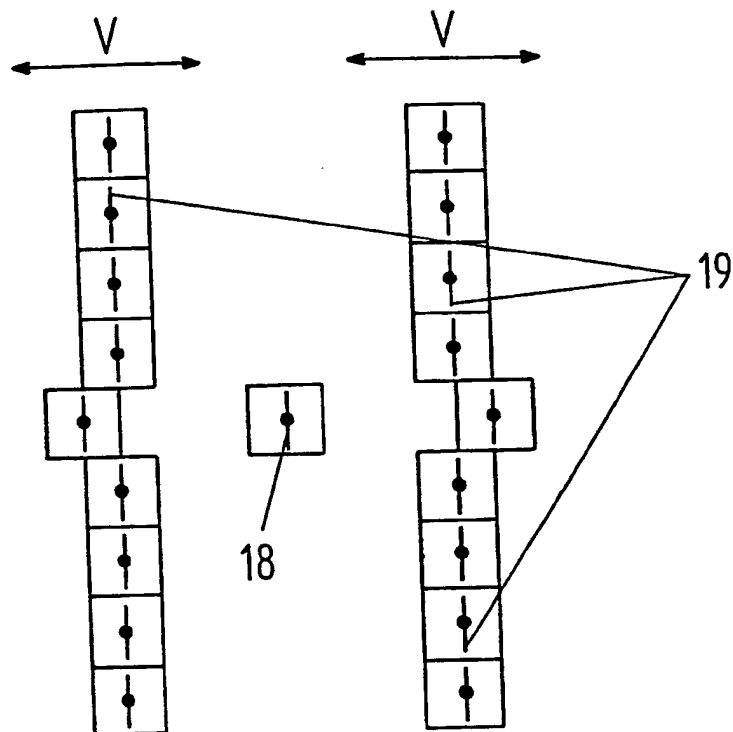


FIG. 3(b)

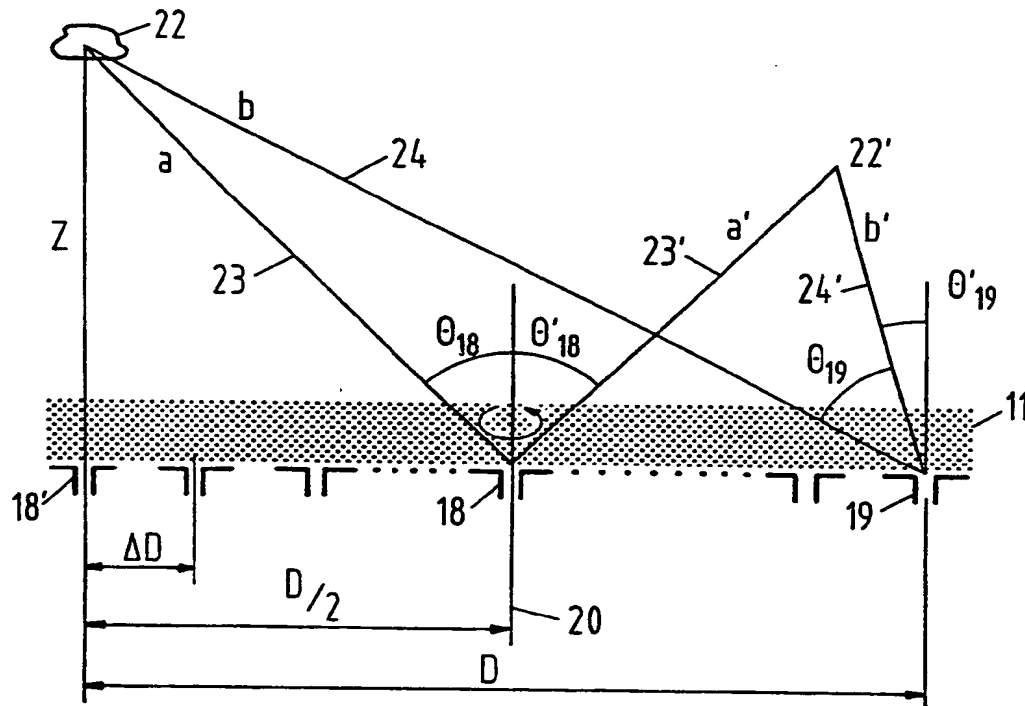


FIG. 4





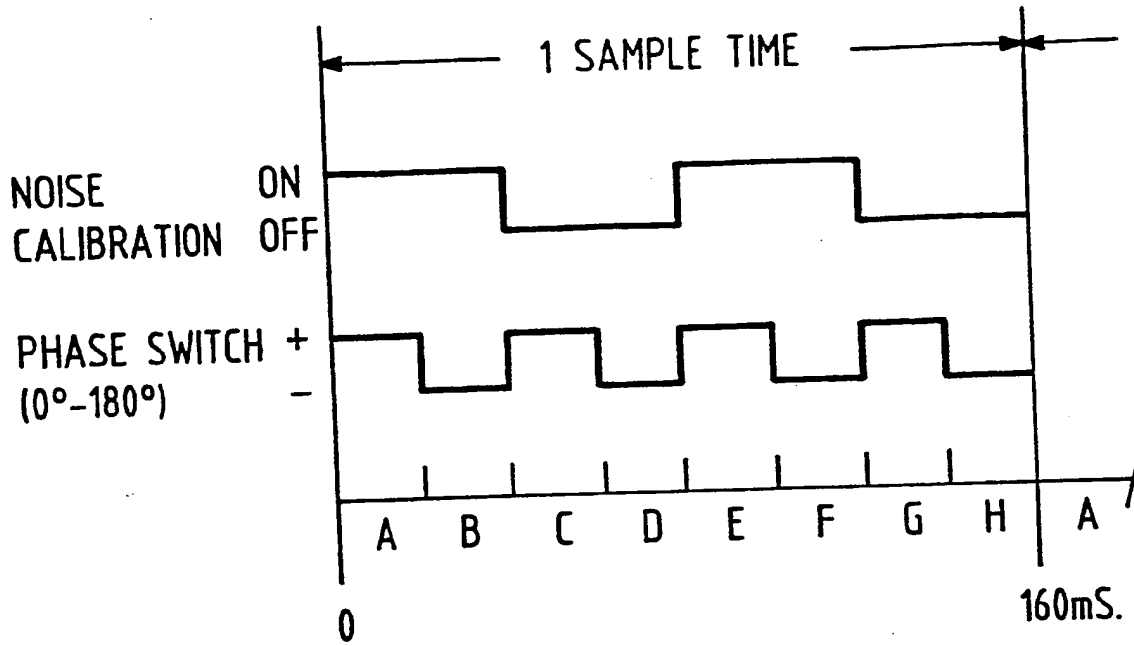


FIG. 7a

TRUE AMPLITUDE AND
PHASE TRACK

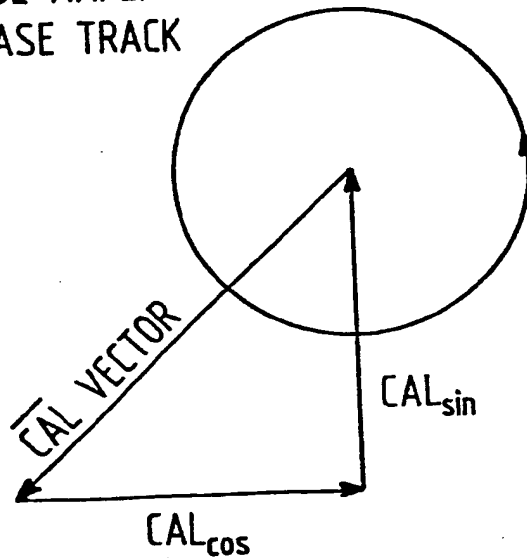


FIG. 7b

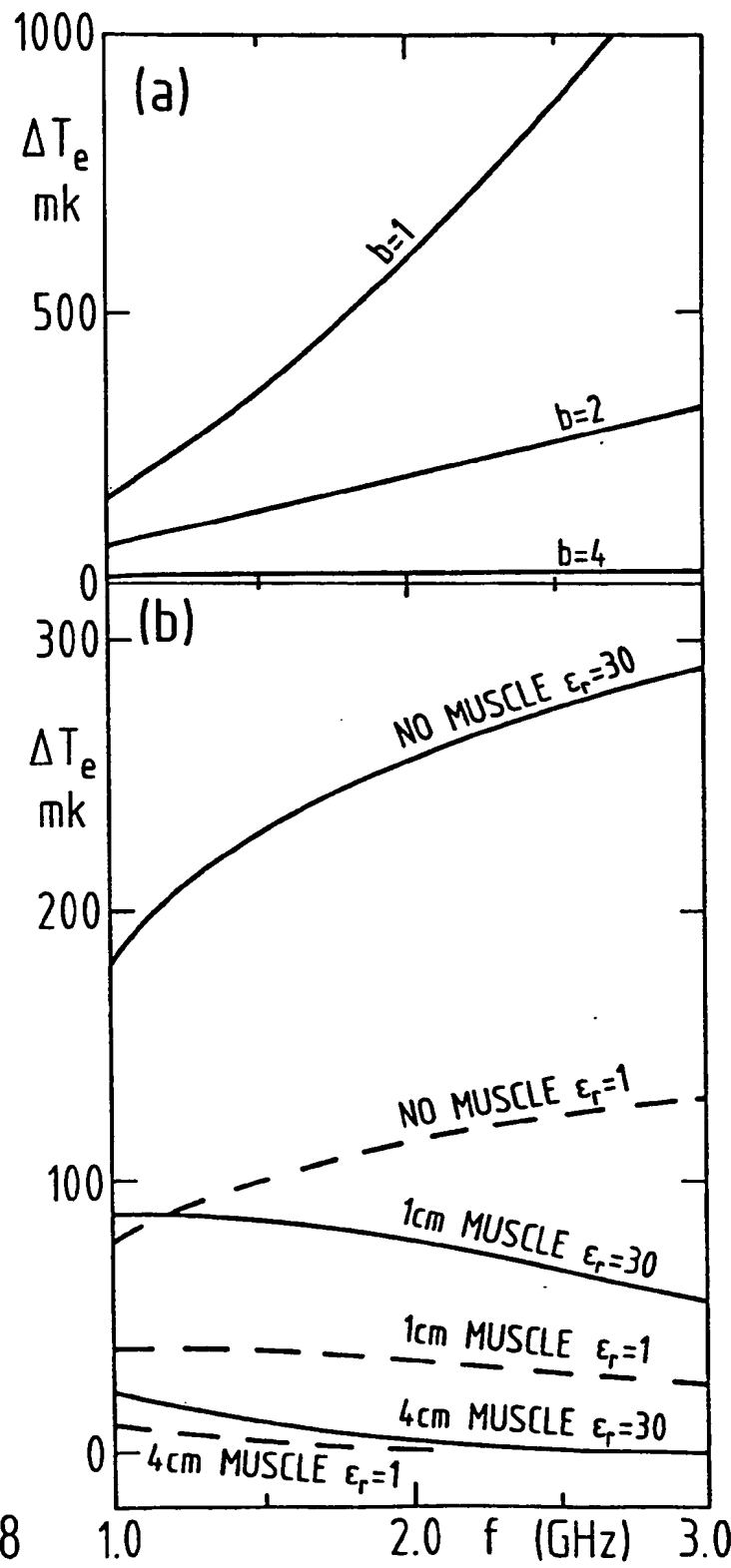


FIG. 8



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EUROPEAN SEARCH REPORT

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Application number

EP 84 10 4365

DOCUMENTS CONSIDERED TO BE RELEVANT			
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl.4)
Y	GB-A-2 052 909 (INSTRUMENTARIUM CY.) * Page 2, lines 10-40; figure 1 *	1	G 01 R 29/08
A		3	
Y	--- CONFERENCE PROCEEDINGS, 12th EUROPEAN MICROWAVE CONFERENCE, 13th-17th September 1982, pages 553-558, Helsinki, FI; A. MAMOUNI et al.: "Introduction to correlation microwave thermography" * Page 554, figure 1; paragraph: "Principle of the correlation microwave thermography" *	1	
Y	--- US-A-4 178 100 (ROBERT A. FROSCH) * Abstract; figure 2 *	1	TECHNICAL FIELDS SEARCHED (Int. Cl.4)
A		6, 12, 15, 16	G 01 R A 61 B A 61 N
A	--- US-A-4 416 552 (ROBERT A. HESSEMER, LLOYD J. PERPER) * Figures 12-16; claims 1, 2, 4-7, 14-15, 18 *	5-9	
--- -/-			
The present search report has been drawn up for all claims			
Place of search THE HAGUE		Date of completion of the search 10-12-1984	Examiner KAUFFMANN J.
CATEGORY OF CITED DOCUMENTS			
X : particularly relevant if taken alone Y : particularly relevant if combined with another document of the same category A : technological background O : non-written disclosure P : intermediate document		T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document	

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Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl. 4)
A	1982 IEEE MTT-S INTERNATIONAL MICROWAVE SYMPOSIUM DIGEST, 15th-17th June 1982, session paper V2, pages 438-440, Dallas, USA; F. STERZER et al.: "A self-balancing microwave radiometer for non-invasively measuring the temperature of subcutaneous tissues during localised hyperthermia treatments of cancer" * Figure 2 *	19,20	
A	IEEE MTT-S INTERNATIONAL MICROWAVE SYMPOSIUM DIGEST 1983, 31st May - 3rd June, Session F4, pages 186-188, Boston, USA; Y. LEROY et al.: "Present results and trends in microwave thermography"		TECHNICAL FIELDS SEARCHED (Int. Cl. 4)
The present search report has been drawn up for all claims			
Place of search THE HAGUE		Date of completion of the search 10-12-1984	Examiner KAUFFMANN J.
CATEGORY OF CITED DOCUMENTS			
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O	non-written disclosure	L	document cited for other reasons
P	intermediate document	&	member of the same patent family, corresponding document

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